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Development of a Low-cost Biomedical Device to Enhance Pneumonia Diagnosis in Children

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Abstract

Pneumonia has contributed greatly to child mortality, especially among children under the ages of five in sub-Saharan Africa, killing more children than the number of children dying from HIV/AIDS. The current methods of diagnosing pneumonia involved physical examination and chest x-ray which are limited by low accuracy, high error margins, higher cost, and stands the risks of inducing cancer. In this work, a low-cost, non-invasive biomedical device was designed and developed to improve accuracy in diagnosing pneumonia. The device functions to detect fluid in a lung consolidated by pneumonia. Dry grouting sponge was used as a phantom for a healthy lung, while a wet sponge was used to mimic a pneumonia-consolidated lung. Surface exciter was used to produce sound waves which travelled through one side of the phantom and are detected on the other end using an electronic stethoscope. The signals detected were digitally analyzed using MATLAB and AUDACITY software. The differences in resonant frequencies from the power spectrum analysis of sound waves as they travelled through the sponges were used to distinguish between a pneumonia-consolidated lung and a healthy lung.

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1.0 INTRODUCTION

Pneumonia is a respiratory infection that causes inflammation of alveoli in the lungs [1]. Alveoli are air sacs known to be responsible for gaseous exchange (oxygen and carbon dioxide) between the lungs and the bloodstream [1,2]. An inflamed alveolus could be filled with fluid or puss which makes it difficult for a patient with pneumonia to breath due to lack of oxygen [3,4]. When there is a lack of oxygen in the alveoli, it could cause other body organs to malfunction. Typical examples are the malfunctioning of the

heart and the liver which eventually leads to death [5,6]. The agents responsible for spreading pneumonia include viruses, bacteria and fungi [3,7].

Pneumonia has been a leading worldwide cause of death among children [7]. Yearly statistics shows that pneumonia kills approximately 1.4 million children under the ages of five, accounting for 18 % of all deaths worldwide [7-9]. Pneumonia is highly prevalent in South Asia and sub-Saharan Africa [10]. It has been reported that, 50 % of deaths occurred in Africa due to pneumonia, with most deaths being concentrated in the sub-Saharan countries such as Uganda and Malawi. In Uganda, pneumonia caused ~6 million deaths per year among children under the ages of five. Also, pneumonia became the single biggest killer disease in Malawi, killing ~1,000 babies and young children [11].

Current clinical methods commonly used to diagnose pneumonia involve a physical examination and chest x-ray [8]. With physical examination (a common practice in Africa), pneumonia has been misdiagnosed for malaria among children in Uganda in most cases. Only 1 in 5 caretakers can diagnose pneumonia correctly with this method [12]. The inability to diagnose the symptoms of pneumonia caused an increase in mortality rates in Uganda [13] when children were sometimes offered antimalarial drugs [14] due to poor diagnosis. This poor treatment results in more death of children, hence a call for a more effective and low-cost method for diagnosing pneumonia in resource-limited underdeveloped areas.

Diagnostic devices for proper detection of pneumonia are characterized by high sensitivity and specificity. Sensitivity signifies the ability of the device to detect pneumonia in a patient with high precision, while specificity implies the inability of the device to identify pneumonia correctly. During physical examinations, doctors observe a patient based on symptoms to detect the possibility of a patient who has pneumonia. The symptoms include high breathing rate and pulse rate. However, the sensitivity of this method is as low as 58 % and its specificity is 67 %. This is because symptoms like difficulties in breathing are also found in other diseases like asthma [9]. Pneumonia and asthma are commonly misdiagnosed since both diseases have symptoms of difficulties in breathing. Only few health facilities in Africa uses chest x-ray for the determination of fluid accumulated in a lung. Its sensitivity is 74 % and specificity is 84 % [15,16]. However, x-ray many not be readily accessible in resource-limited underdeveloped areas. X-ray works through the release of electromagnetic radiation into the body of a patient. However, it is not recommended for a person to undergo an x-ray examination more than once a year since it can increase the chances of getting cancer [10], [17].

Several studies which include automated and non-automated methods have been explored to diagnose pneumonia based on pleural fluid and lung sounds [11], [18]. The use of physical methods in the detection of pneumonia includes the chest auscultatory percussion where a sound stimulus is introduced to the chest of a patient through tapping the chest with a finger and detecting the audio change at the back of the chest using a stethoscope [19-21]. This method is based on sound transmission and distortion when sound waves travel in different mediums and can be used to detect if a lung is filled with fluid (that is, air or liquid) [22]. Previous studies also developed a computational model whereby sound waves were transmitted through the chest. It explored the use of lung acoustics (electromagnetic shaker) which was then introduced on the sternum to generate an automated sound input. A laser doppler vibrometer was then used to measure the vibrations at the back [23,24]. The system was used in the detection of pneumothorax which is a condition in which the lung tissues collapse due to the accumulation of air in the spaces between lungs (pleural space) [24,25]. This model is proven to be effective but had low portability since the input device was heavy (> 2 kg) which maybe expensive for resource-limited underdeveloped areas.

The current work, therefore, presents a device that is relatively portable and low-cost (affordable) for resource-limited underdeveloped settings.

Moreover, electronically automated methods prove to be more reliable since digital information is free from interobserver errors and allow for reproductively in results. However, there have not been many efforts towards the use of digitized tools for accurate diagnosis of pneumonia. Part of this will be addressed in this paper towards providing effective detection and diagnosis of pneumonia. The goal of this work is to design a low-cost, non-invasive diagnosis system that can increase detection sensitivity, accuracy and specificity during pneumonia detection. Charges on pneumonia diagnosis via chest x-ray cost about USD 200-400 which is quite expensive for the low-income earning citizens. However, the cost of building the current system is as low as USD 150.

2.0 MATERIALS AND METHODS

2.1 Materials

The software, Audacity and MATLAB were acquired by the Ashesi University. A surface exciter (HYD-20) and the grouting sponge (QEP 70005Q-6D) were purchased from Amazon. The Arduino microcontroller, capacitors, resistors, and amplifiers were provided by the Department of Engineering, Ashesi University.

2.2 Block Diagram of Device

The summary of the operation of the device is shown in the block diagram (Fig. 1). From Figure 1, a battery was used to power the electronic circuit. The surface exciter produces and then transmits sound waves through one side of the lung phantom (grouting sponge mimicking a human lung). The sound waves were then detected by a stethoscope at the other end of the lung phantom. The sound waves were then analyzed using MATLAB and Audacity software and the results underwent digital signal processing. Resonant frequencies of the sound waves passing through the lung phantoms were determined from a spectrum analysis.

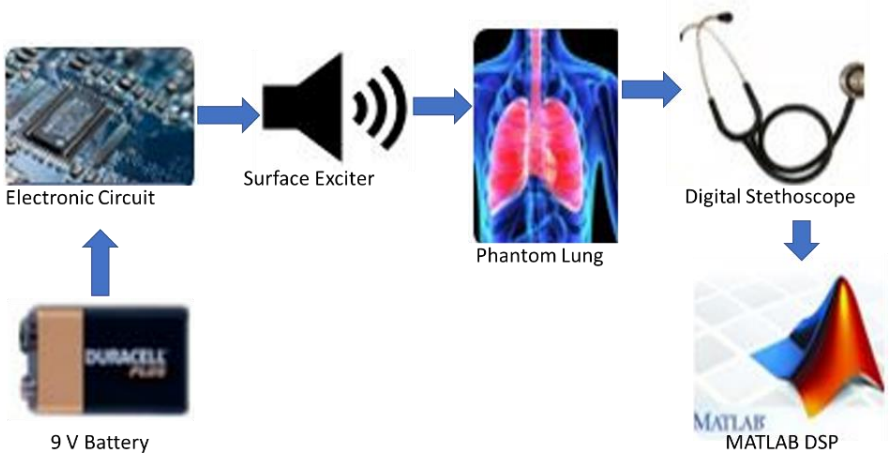

Figure 1: Block Diagram of Device.

2.3 Actuator Selection

An actuator was chosen based on its size, the frequency range of sound waves it produces, and the intensity of the sound waves at low frequency. Common actuators were considered. The following Pugh chart in Table 1 was developed for the actuator selection. A scale of 1 to 5 was used where '1' represents a low response and '5' represents a high response. The weights of the criteria were extracted from the design standards set by UNICEF [26].

Table 1: Pugh Chart for Actuator.

Criteria	Weight	Push/Pull Solenoid	Piezoelectric Transducer	Speaker	Surface Exciter
Intensity	5	5	1	2	4
Cost	3	3	4	3	3
Portability	3	1	5	4	4
Battery Life	2	2	4	4	4
Total Score		41	40	39	49

[1]*low actuator performance and* [5]*high actuator performance.*

2.3 Lung Phantom

Proteus software was used to design the digital electronic stethoscope circuit (Fig. 2). A 9 V power supply was used to power the circuit. Amplifier U1:A (NE5532) is a dual, low-noise amplifier. It represents the signal acquisition circuit which consists of the electret microphone connected to a pull-up resistor, R_1. The electret microphone is a transducer that converts sound signals to electrical signals that can be amplified and filtered. Capacitor, C_1, removes the direct current (dc) offset hence, allowing only alternating current (ac) output. The circuit makes use of the negative feedback system to generate and amplify the ac output of the microphone and bring it to a suitable level. The electret microphone is attached to a chest piece (taken from a manufactured stethoscope) to insulate the microphone from background noise when the stethoscope head is placed at the surface of the sponge.

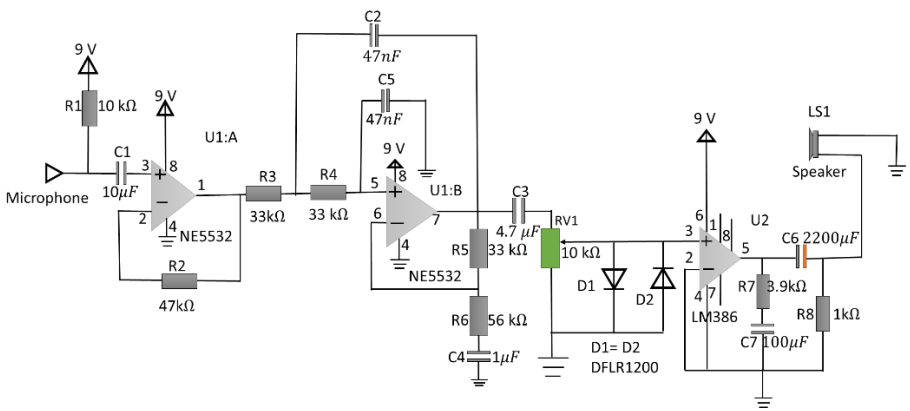

Figure 2: Electronic Stethoscope for Lung Sound Detection.

The amplified signal has noise components, hence, the need to filter and remove the noise. Amplifier U1:B represents a low noise Sallen-Key Butterworth low-pass filter. The cut-off frequency of this filter is calculated using the following equation.

$$F_c = \frac{1}{2\pi RC} = \frac{1}{2\pi(33 * 10^3)(47 * 10^{-9})} = 102.6\ Hz \quad (1)$$

where R is the resistance of the resistor and C is the capacitance of the capacitor. The transfer function for the Sallen-Key lowpass filter was determined using the following equations (2a and 2b) [27, 28]. The Sallen-Key filter allows only signals with frequencies below 102.3 Hz and cuts off the signals with frequencies above the cut-off frequency. The transfer function for the Sallen-Key lowpass filter was determined using the following equations (2a and 2b) [27,28].

$$G(s) = \frac{K}{R_3R_4C_2C_5s^2 + s(R_3C_2 + R_4C_5 + R_3C_2(1-K) + 1} \quad (2a)$$

where K is a constant (= 1), the R's are resistors, the C's are capacitors and s is a variable of the frequency domain.

$$G(s) = \frac{416666}{s^2 + 1292s + 416666} \quad (2b)$$

After filtering, the capacitor, C_3, blocks the dc from passing through, hence only allowing the ac signal. This ac signal was amplified by the audio amplifier, LM386. Capacitor, C_6, filtered the dc signal and the ac output signal was then sent to the speaker using an aux cable to ensure that the sound can be heard on the speaker. The microphone was also separated from the speaker to prevent the development of acoustical feedback which distorts the results. The full breadboard connection of the device is as shown below in Figure 3.

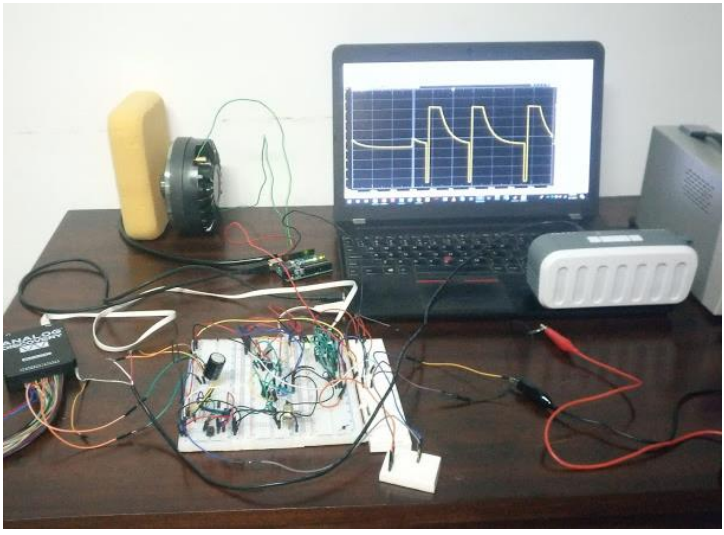

Figure 3: Device Full Breadboard Connection.

The power spectrums of the sound waves as they pass through each sponge were plotted and the resonant frequencies determined. The differences in the resonance frequencies were used to categorize a pneumonia-infected lung and a healthy lung.

3.0 RESULTS AND DISCUSSION

The signal profiles as the sound travelled through the sponges were recorded using Audacity software (Fig. 4). The processes of noise removal, normalization, equalization, and compression were done to filter and amplify the signal. Normalization was done to bring the signal profile to a target level. A gain of -2 dB was used during normalization. Equalization of the signal was done to boost the frequencies of the signal to bring out a more natural sound. The resulting signal profiles after filtering and amplifying are shown in Figure 4 below.

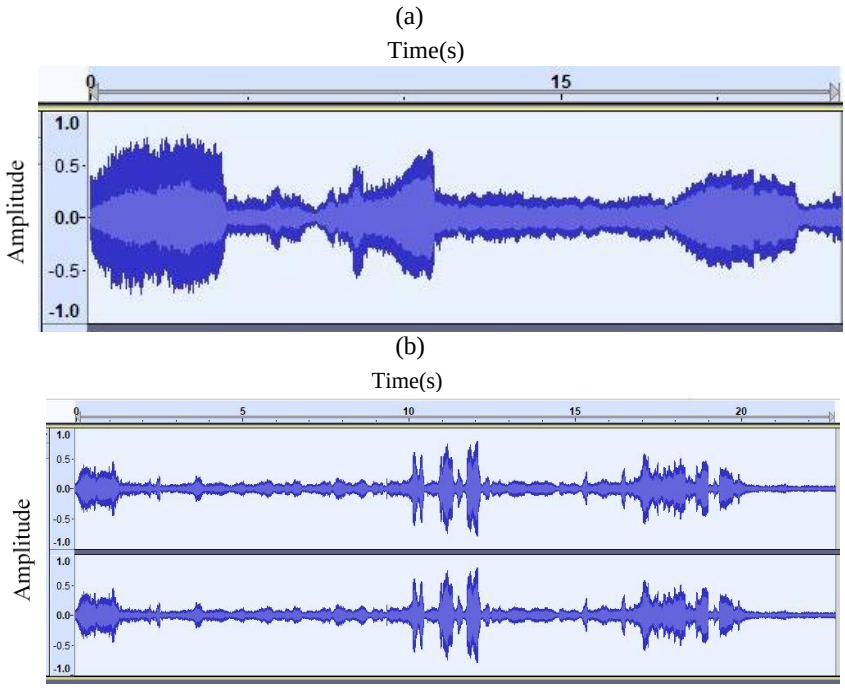

Figure 4: (a) Raw Signal after Filtering and Amplification for the Dry Sponge.

The spectrogram is a qualitative analysis that shows the variation of frequency with time. The spectrogram for signals that pass through the dry sponge and wet sponge is as shown (Figs. 5a-b). The different colors showed the spread of the frequency of the signal as it passes through the sponge. More frequencies of the signal are concentrated in the region less than 5 seconds for the dry sponge (Fig. 5a). The frequencies of the signal passing through the wet sponge are fairly distributed over time (Fig. 5b). The amplitudes are very low which implies that the wet sponge absorbs most of the signals.

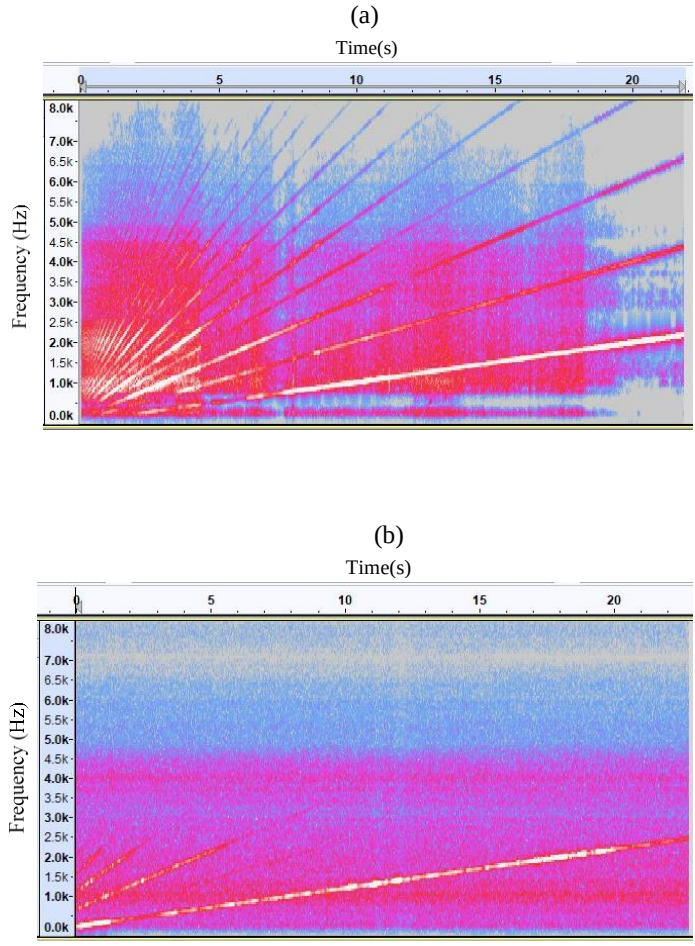

Figure 5: Spectrum of Sound Waves out of the: (a) Dry Sponge and (b) Wet Sponge.

In other to determine the quantitative analysis of the sound waves as they pass through the dry sponge, a spectrum of the signal for each of the sponges was determined. The power spectrums were then plotted with OriginPro-8, and the result is shown in Figure 6.

The resonance frequency shows the strength of the sound waves propagating through the lung phantom. The resonance frequency is the determined power spectrum obtained from the Audacity software. The software has a built-in algorithm for calculating and plotting power spectrum graphs (Fig. 6).

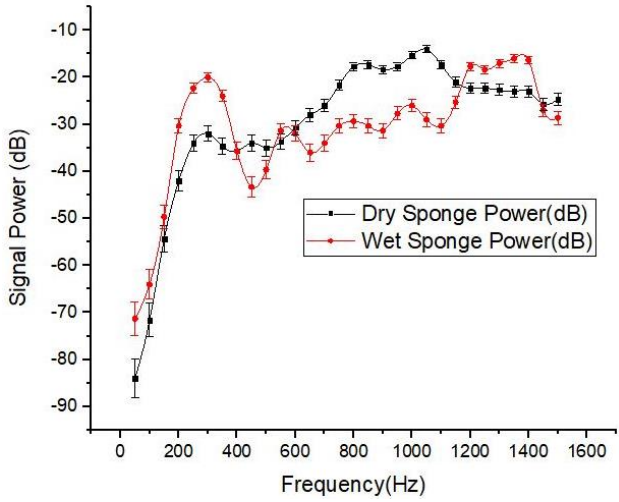

Figure 6: Signal Power (dB) for Both Dry and Wet Sponge at Variable Frequency.

For the dry sponge, the resonant frequency was recorded at 1050 Hz (at sound waves of -14 dB), while the resonant frequency for the wet sponge was 300 Hz (at -20 dB). At the resonant frequency of the wet sponge (300 Hz), the wet sponge acts as a high pass filter (allowing fewer signals above 300 Hz to pass through), while the dry sponge acts as a high pass filter (allowing more signal above the 300 Hz to pass through the sponge). This difference in the behavior of the signal power can be used in the diagnosis of pneumonia. Sound waves traveling through a pneumonia-consolidated lung (having fluid) have a resonant frequency of less than 500 Hz, while sound waves traveling through a pneumonia free lung (healthy lung) have a resonant frequency of around 1000 Hz.

Although the most important parts of the device including the stethoscope were working, the electret microphone was not very sensitive enough to detect low signals. The use of a more sensitive already-made stethoscope is needed to improve the sensitivity of the device to detect very low signals. The implications of this work for the rapid diagnosis of pneumonia in low-resource settings are that instead of relying on physical examination method which has high error margin, the diagnosis of pneumonia can be improved to a better accuracy and at a low cost. A correct diagnosis implies proper prescription of drugs. This will help reduce mortality of children who dies through to poor diagnosis.

3.0 CONCLUSIONS

This work shows a distinct difference in the resonant frequencies when sound waves pass through wet and dry sponges. The work shows that sound behaves differently when it passes through different mediums. The results also correlate with physical theories of sound transmission in different mediums, that sound travels faster in liquids than in gases. With a pneumonia consolidated lung being filled with fluid, the device can be used to detect the fluid in children under the ages of five by checking the power of the signal as it travels through the chest of the child. Thus, the differences in resonant frequencies from the power spectrum of sound waves as they travelled through the

sponges were used to distinguish between a pneumonia-consolidated lung and a healthy lung.

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